SPECIFICATION

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METHOD FOR DYNAMIC STABILIZATION OF PET DETECTOR GAINS

Background of the Invention

[0001] The present invention relates to PET scanners generally and specifically to a method and apparatus for adjusting PMT gains to compensate for drift due to various operating phenomenon.

[0002] Positrons are positively charged electrons which are emitted by radionuclides which have been prepared using a cyclotron or other device. The radionuclides most often employed in diagnostic imaging are fluorine-18, carbon-11, nitrogen-13 and oxygen-15. Radionuclides are employed as radioactive tracers called "radiopharmaceuticals" by incorporating them into substances such as glucose or carbon dioxide. One common use for radiopharmaceuticals is in the medical imaging field.

[0003] To use a radiopharmaceutical in imaging, the radiopharmaceutical is injected into a patient and accumulates in an organ, vessel or the like, which is to be imaged. It is known that specific radiopharmaceuticals become concentrated within certain organs or, in the case of a vessel, that specific radiopharmeceuticals will not be absorbed by a vessel wall. The process of concentrating often involves processes such as glucose metabolism, fatty acid metabolism and protein synthesis. Hereinafter, in the interest of simplifying this explanation, an organ to be imaged will be referred to generally as an "organ of interest" and prior art and the invention will be described with respect to a hypothetical organ of interest.

[0004]

After a radiopharmaceutical becomes concentrated within an organ of interest and

while the radionuclides decay, the radionuclides emit positrons. Each positron travels a very short distance before it encounters an electron and, when the positron encounters an electron, the positron is annihilated and converted into two photons, or gamma rays. This annihilation event is characterized by two features which are pertinent to medical imaging and particularly to medical imaging using photon emission tomography (PET). First, each gamma ray has an energy of essentially 511 keV upon annihilation. Second, the two gamma rays are directed in substantially opposite directions.

[0005] In PET imaging, if the general locations of annihilations can be identified in three dimensions, the shape of an organ of interest can be reconstructed for observation. To detect annihilation locations, a PET scanner is employed. An exemplary PET scanner includes one or more rings of detector modules and a processor which, among other things, includes coincidence detection circuitry. The detector modules are arranged about an imaging area. An exemplary detector module includes six adjacent detector blocks. An exemplary detector block includes an array of 36 bismuth germinate (BGO) scintillation crystals arranged in a 6X6 matrix and four photo-multiplier tubes (PMTs) arranged in a 2X2 matrix to the side of the crystal matrix opposite an imaging area.

[0006] When a photon impacts a crystal, the crystal generates light which is detected by the PMTs. The PMT signal intensities are combined to generate a combined analog signal which is converted into a digital signal. For the purposes of this explanation, it will be assumed that the digital value, also referred to as a target value, to 511 keV is 180. The combined digital signal is compared to a range of values about 511 keV. When the combined signal is within the range, an event detection pulse (EDP) is generated which is provided to the processor coincidence circuitry. In addition, acquisition circuits determine which crystal within a block absorbed the photon by comparing the relative strengths of the PMT signals.

[0007] The coincidence circuitry identifies essentially simultaneous EDP pairs which correspond to crystals which are generally on opposite sides of the imaging area. Thus, a simultaneous pulse pair indicates that an annihilation has occurred somewhere on a straight line between an associated pair of crystals. Over an

acquisition period of a few minutes, millions of annihilations are recorded, each annihilation associated with a unique crystal pair. After an acquisition period, recorded annihilation data is used via any of several different well known procedures to construct a three dimensional image of the organ of interest.

[0008] While operation of a PET detector is relatively simple in theory, unfortunately, despite efforts to manufacture components that operate in an ideal fashion, there is an appreciable variation in how similar detector components respond to identical stimuli. For example, given a detector block including 36 crystals and four PMTs and given the same stimuli, crystals that are positioned proximate the center of the PMT array will typically generate a higher energy value than edge or corner crystals (i.e., crystals that are positioned along the edge of the array or at the corner of the array). This disparate and position dependent energy spectrum occurs because, typically, some of the light generated by an edge or corner crystal is not detected by the PMTs in a single block.

[0009] As one other example, even within a single crystal, impacting photons may not generate the same PMT output for various reasons. For instance, some photons are completely absorbed by a crystal while others are not. Completely absorbed photons generate light corresponding to 511 keV while partially absorbed photons generate less than the 511 keV. As another instance, first and second photons may be partially absorbed essentially simultaneously by first and second crystals in the same block. While each photon would be identified if they had been absorbed consecutively, upon simultaneous absorption, the combined energy may erroneously be attributed to a single absorbed photon. In this case detection circuitry may erroneously identify a third crystal between the first and second crystals as the detecting or absorbing crystal.

[0010] Thus, while each detected photon should ideally generate a signal having an energy level of 511 keV, in many cases detected photons generate much less energy. For this reason, the energy range used to determine if a combined digital PMT signal corresponds to a detected photon typically is assigned a relatively low threshold value. For instance, in an exemplary PET system the low end of the energy range may be a digital value of 35 corresponding to approximately 100 keV (i.e., any absorption even

having an energy greater than 100 keV is assumed to correspond to a photon).

[0011] In addition to the potential errors described above, other sources of system error also occur. For instance, given two PMTs and identical stimuli (i.e., input light), a first PMT will typically generate a slightly different output signal than the second PMT. Exacerbating matters, over time PMT performance has been known to degrade due to aging related changes in structure. Further exacerbating matters, PMTs often operate differently when exposed to different operating parameters. For instance, PMT output signals have been known to vary as a function of temperature, ambient magnetic fields and other parameters that are relatively difficult or expensive to control.

[0012] To compensate for PMT construction and operating variances, the PET industry has developed various commissioning/calibration procedures and associated hardware and software. Generally, during a calibration procedure, a PET source having a known intensity is provided inside the PET imaging area and PMT signals generated thereby are collected. The collected PMT signals are compared to expected PMT signals and, where there is a difference between the collected and known signals, PMT gains are adjusted to compensate for the differences.

[0013] While calibration techniques like the one described above are useful, unfortunately, most calibration techniques require acquisition of massive amounts of data and hence an appreciable amount of time to complete. In addition, many calibration techniques include at least some manual steps that have to be performed by skilled technicians.

[0014] Because of the time and skills required to calibrate a PET system, in many cases, calibration will only be performed when absolutely necessary such as after image artifacts begin to appear in generated images. In other cases calibration is performed routinely whether or not the calibration is necessary. For instance, in some cases calibration is performed on a weekly basis. In the case of mobile PET systems (i.e., truck based systems), the system environment and, in particular, ambient magnetic fields, may change on a daily basis. In these cases calibration will typically be performed on a daily basis.

[0015] Thus, in some cases where calibration should be performed, calibration may be

foregone until a later time while in other cases, where calibration is not necessary, a routine calibration procedure may be performed. In addition, in cases where calibration is only performed when a radiologist begins to recognize artifacts, the radiologist is routinely faced with the question of whether or not to recalibrate.

Brief Summary of the Invention

[0016] The above discussed and other drawbacks and deficiencies are overcome or alleviated by a method for calibrating PET detector PMT gains in a detector unit, the method including the steps of conducting a calibration procedure, determining whether a number of photons absorbed by a corresponding crystal exceeds a count threshold, if the threshold is exceeded, repeating the steps of conducting a calibration procedure and determining whether a number of photons absorbed by a corresponding crystal exceeds a count threshold, and if the threshold is not exceeded, ending the calibration procedure.

[0017] In an alternative embodiment, an apparatus for calibrating PET detector PMT gains in a detector unit, the apparatus including means for conducting a calibration procedure, means for determining whether a number of photons absorbed by a corresponding crystal exceeds a count threshold, means for repeating the calibration procedure if the threshold is exceeded, and means for ending the calibration procedure if the threshold is not exceeded.

[0018] In an alternative embodiment, an apparatus for calibrating PET detector PMT gains in a detector unit, the apparatus including a processor for performing a pulse sequencing program to perform the steps of conducting a calibration procedure, determining whether a number of photons absorbed by a corresponding crystal exceeds a count threshold, wherein if the threshold is exceeded, repeating the steps of conducting a calibration procedure and determining whether a number of photons absorbed by a corresponding crystal exceeds a count threshold, and further wherein if the threshold is not exceeded, ending the calibration procedure.

[0019]

In an alternative embodiment, a method for improving image performance in PET imaging by stabilizing gain in compensators which are separately adjustable so that received digital signals are adjusted to compensate for PMT degradation, includes

calculating PMT signal adjustments within a calibrator and adjusting output of compensators with the PMT signal adjustments, wherein the steps of calculating and adjusting are repeated during image acquisition in a PET scanner system until a number of photons absorbed by a corresponding crystal does not exceed a count threshold.

[0020] In an alternative embodiment, an imaging system includes a scanner system, an image reconstruction processor, and ALC circuitry, wherein the ALC circuitry includes a calibrator for calculating gain adjustment of compensators within the ALC circuitry and further wherein the image reconstruction processor includes program signals for defining an executable program for repeating calculation of gain adjustment in the calibrator during image acquisition performed by the scanner system.

[0021] The above discussed and other features and advantages of the present invention will be appreciated and understood by those skilled in the art from the following detailed description and drawings.

Brief Description of the Drawings

[0022] Fig. 1 is a schematic view of a PET system for implementing the present invention;

[0023] Fig. 2 is a perspective view of a detector unit and associated PMTs.

[0024] Fig. 3 is a schematic view of the ALC circuitry of Fig. 1;

[0025] Fig. 4 is a flow chart illustrating a commissioning procedure;

[0026] Fig. 5 is a flow chart illustrating a calibration procedure;

[0027] Fig. 6 is a flow chart illustrating an alternate calibration procedure;

[0028] Fig. 7 is a graph illustrating a crystal spectrum generated during a commissioning procedure; and

[0029] Fig. 8 is a graph illustrating a combined unit spectrum generated during a calibration procedure.

Detailed Description of the Invention

[0030] While various components are described below for carrying out several inventive methods, it should be appreciated that all of the methods herein may be performed by any of several different commercially available and programmable processors.

[0031] Referring now to the drawings, wherein like reference characters and symbols represent corresponding elements and signals throughout the several views, and more specifically referring to Fig. 1, the present invention will be described in the context of an exemplary PET scanner system 8. System 8 includes an acquisition system 10, an operator work station 15, acquisition, locator and coincidence (ALC) circuitry 30 and an image reconstruction processor 40.

[0032] System 10 includes a gantry 9 which supports a detector ring assembly 11 about a central bore which defines an imaging area 12. A patient table (not illustrated) is positioned in front of gantry 9 and is aligned with imaging area 12. A patient table controller (not shown) moves a table bed (not shown) into imaging area 12 in response to commands received from work station 15 through a serial communications link 18.

[0033] A gantry controller 17 is mounted within gantry 9 and is responsive to commands received from operator work station 15 through link 18 to operate gantry 9. For example, gantry 9 can be tilted away from vertical on command from an operator, can perform a "transmission scan" with a calibrated radio nuclide source to acquire attenuation measurements, can perform a "coincidence timing calibration scan" to acquire corrective data, or can perform a normal "emission scan" in which positron annihilation events are counted.

[0034] As shown best in Fig. 2, assembly 11 is comprised of a large number of detector blocks 20. Although not illustrated, detector blocks 20 are arranged in modules, each module including six separate and adjacent detector blocks 20. A typical assembly 11 includes 56 separate modules such that each assembly 11 includes 336 separate detector blocks 20. Each block 20 includes a set of bismuth germinate (BGO) scintillator crystals 21 (two separate crystals identified by numerals 180 and 182) arranged in a 6 x 6 matrix and disposed in front of four photo multiplier tubes (PMTs) A, B, C and D which are collectively referred to by numeral 22. When a photon impacts a crystal 21, a scintillation event occurs and the crystal generates light which is

directed at PMTs 22. Each PMT 22 receives at least some light generated by the scintillation event and produces an analog signal 23A-23D which arises sharply when a scintillation event occurs and then tails off exponentially with a time constant of approximately 300 nanoseconds. The relative magnitudes of the analog signals 23A-23D are determined by the position in the 6 x 6 BGO matrix at which a scintillation event takes place, and the total magnitude of these signals is determined by the energy of an absorbed or partially absorbed photon which causes the event.

[0035] Referring still to FIGS. 1 and 2, a set of acquisition circuitry 25 is mounted within gantry 9 to receive the four signals 23A-23D from each detector block 20 in assembly 11. Circuitry 25 provides signals 23A-23D to ALC circuitry 30 via a data bus 26. Circuitry 30 uses the signals 23A-23D to determine the energy E_i of a detected event, if the energy detected likely corresponds to a photon, the actual coordinates C_i of a detected event within the block of BGO crystals 21, the time T_i of the event (i.e. generates a time stamp) and compares event times to identify coincidence pairs of events that are stored as coincidence data packets. Each coincidence data packet includes a pair of digital numbers which precisely identify the addresses of the two BGO crystals 21 that detected an associated event. Operation of ALC circuitry 30 is explained more in detail below.

[0036] Referring again to Fig. 1, processor 40 includes a sorter 34, a memory module 43, an array processor 45, an image CPU 42 and a backplane bus 41 which conforms to the VME standards and links all other processor components together. The primary purpose of sorter 34 is to generate memory addresses for the coincidence data packets to efficiently store coincidence data. The set of all projection rays that point in the same direction and pass through the scanner's FOV is a complete projection, or "view". A distance R between a particular projection ray and a center of the FOV locates that projection ray within the FOV. As shown in Fig. 1, for example, a positron annihilation (hereinafter an "event") 50' occurs along a projection ray 51' which is located in a view at the projection angle θ and the distance R. Sorter 34 counts all of the events which occur on this projection ray (R, θ) during an acquisition period by sorting out the coincidence data packets that indicate an event at the two BGO detector crystals lying on ray 51'.

[0037] During data acquisition, the coincidence counts are organized in memory 43 as a set of two-dimensional arrays, one for each axial image, and each having as one of its dimensions the projection angle θ and the other dimension distance R. This θ by R histogram of detected events is called a sinogram. Coincidence events occur at random and sorter 34 quickly determines the θ and R values from the two crystal addresses in each coincidence data packet and increments the count of the corresponding sinogram array element. At the completion of an acquisition period, memory 43 stores the total number of annihilation events which occurred along each ray (R, θ) in the sinogram.

[0038] Image CPU 42 controls bus 41 and links processor 40 to local network 18. Array processor 45 also connects to the bus 41 and operates under the direction of image CPU 42 to facilitate image reconstruction using histogram data from memory module 43. The resulting image array is stored in memory module 43 and may be output by image CPU 42 to operator work station 15.

[0039] Station 15 includes a CPU 50, a CRT display 51 and a keyboard 52 or other similar input device (e.g., mouse, joystick, voice recognition module, etc.). CPU 50 connects to network 18 and scans key board 52 for input information. Through the keyboard 52 and associated control panel switches, an operator can control calibration of system 9, its configuration, and the positioning of a patient table during data acquisition.

[0040]

Referring still to Figs. 1 and 2 and also to Fig. 3, among other components for each block 20, exemplary and simplified ALC circuitry 30 includes an analog to digital (AD) converter 95, four compensators 70A, 70B, 70C and 70D, an energy, time and crystal identifier 72, coincidence detection circuitry 90 and event discriminator 91. A separate line (collectively identified by numeral 26) links each of PMTs 22 in a "unit" to the AD converter 95 which converts each of the analog signals 23A-23D to a digital signal 23Ad-23Dd. Consistent with the explanation and assumptions above, an analog signal corresponding to 511 keV is converted to a digital value 180, an analog signal corresponding to 100 keV is converted to a digital value 35, etc. Digital values 23Ad-23Dd are provided to compensators 70A-70D, respectively. Each compensator 70A-70D is separately adjustable so that the received digital signal (e.g., 23A) may be

either increased or decreased to compensate for PMT degradation or varying operation due to ambient changes.

[0041] Compensator outputs are provided to identifier 72 which uses the received compensated PMT signals to identify event energy levels E_i, time T_i at which each event occurs and which crystal C_i absorbed the photon that caused the event. Methods and circuitry to perform each of these tasks are well known in the PET industry and therefore will not be explained here in detail. For a better understanding of how identifier 72 operates refer to U.S. Pat. No. 6,232,604 which is entitled Analog Time Adjustment for Coincidence Detection Electronics.

[0042] Referring still to Figs. 1 through 3, the event times T_i, energies E_i and crystal identifiers C_i are provided to event discriminator 91 as distinct data packets. Discriminator 91 compares the event energies E_i to a threshold energy level E_{th} to identify data packets that likely correspond to valid absorbed photons. In the present example, it will be assumed that energy level E_{th} is 100 keV corresponding to a digital value 35. Thus, discriminator identifies all packets having digital values between 35 (e.g., 100 keV) and 180 (e.g., 511 keV) and passes times T_i and identifiers C_i corresponding to those packets to coincidence detection circuitry 90.

[0043] Circuitry 90 compares the times T_i in each packet to identify coincidence events. Where two even times T_i are within a small period (i.e., within a "coincidence window") and if other criteria (e.g., corresponding crystals C_i are separated by the system field of view (FOV)) are met, circuitry 90 identifies the packets as comprising to a "coincidence pair." Circuit 90 provides coincidence pairs to sorter 34 for further image processing as described above.

[0044] Hereinafter, in the interest of simplifying this explanation the term "unit" will be used to refer to 25 blocks 20. However, it should be appreciated that the invention contemplates other groupings of blocks. For instance, in some cases a unit may include two blocks 20, four blocks 20, all blocks 20 that reside in an upper half of detector 11, all blocks that reside in the lower half of detector 11, all blocks 20 within detector 11, etc. the smallest unit comprises a single block 20.

[0045]

Referring to Fig. 3, ALC 30 also includes one or more calibrators 69. During each

of a commissioning and a calibration process, calibrator 69 receives all energy E_i and crystal identifier C_i information from identifiers 73 (i.e., there is one identifier 72 for each block) corresponding to a single unit. Using the received information calibrator 69 calculates PMT signal adjustments used to adjust the output of compensators 70A-70D. Thus, in the present example, calibrator 69 receives signals from 25 separate blocks 20 (see also Figs. 1 and 2).

[0046] Referring still to Fig. 3, calibrator 69 is used to perform two separate processes. A first process, referred to as a commissioning process, is to be performed once or perhaps very seldom (e.g., every month or quarter) to generate information that can be used subsequently during calibration processes. The second process, referred to as a calibration process, is meant to be performed routinely. For example, because the calibration process requires only minimal time to complete, it is contemplated that a PET system could perform the calibration process before every data acquisition procedure and in a manner that is imperceptible to both patient and operator. Alternatively, the calibration process may be performed repeatedly, even during an image acquisition procedure. An exemplary commissioning procedure 92 is illustrated in Fig. 4 while an exemplary calibration procedure 114 is illustrated in Fig. 5, and an alternative exemplary calibration procedure 140 is illustrated in Fig. 6.

[0047] Referring still to Fig. 3, calibrator 69 includes a spectrum generator 74, a switch 76, a first peak identifier 78, a gain factor determiner 80, a spectrum shifter 82, a spectrum normalizer 84, a second peak identifier 86, a comparator 88 and an adjuster 89. While shown separately to simplify this explanation, identifiers 78 and 86 may comprise a single identifier. Generator 74 and switch 76 are used during both the commissioning and calibrating process, identifier 78 and determiner 80 are used during the commissioning process and shifter 82, normalizer 84, identifier 86, comparator 88 and adjuster 89 are used during the calibration process.

[0048]

Referring to Figs. 1, 3 and 4, at block 94 a commissioning photon source is provided within imaging area 12 adjacent detector blocks 20. The source (not illustrated) directs photons at blocks 20. For each absorbed photon, identifier 72 provides both the energy E_i and the crystal identifier C_i to spectrum generator 74. Throughout a commissioning period, generator 74 generates a separate

commissioning energy spectrum for each crystal within the unit. Thus, for instance, referring again to Fig. 2, because a unit includes twenty-five blocks 20 in the present example and each block 20 includes 36 crystals (e.g., 180, 182, etc.), generator 74 generates 900 separate crystal spectrums for the exemplary unit.

[0049] Referring now to Fig. 6, an exemplary commissioning spectrum 130 for a single crystal is illustrated. Referring also to Fig. 2, it will be assumed spectrum 130 corresponds to edge crystal 182. Each spectrum 130 plots the number of photons absorbed by a corresponding crystal at specific energy levels on a vertical axis against energy level (i.e., E_i) on a horizontal axis. For instance, in exemplary spectrum 130 in Fig. 6, approximately five thousand absorbed photons had energy levels corresponding to a digital count of 80, approximately twenty thousand absorbed photons had energy levels corresponding to a count of 120 and approximately two thousand photons had energy levels corresponding to a count of 148. To generate a spectrum 130 generator 74 simply maintains counters for each possible digital energy value for each crystal and increments the appropriate counter when a photon energy level E_i matches the level associated with the counter.

[0050] Exemplary commissioning spectrum 130 clearly illustrates that the energies E_i attributed to separate absorbed photons vary widely even within a given crystal due to phenomenon described above including partial absorption, dual absorption, partial light detection, etc. Clearly there is a peak energy level E_p at which a curve through the count values is at a highest point. In spectrum 130 the peak energy level occurs at approximately 115.

[0051] As indicated above, the digital value attributable to a completely absorbed and detected photon is 180 (corresponding to 511 keV). Given this assumption, the peak energy level 115 seams to be relatively low as one would expect the peak level to have been approximately 180. As it turns, crystal 182 (see Fig. 2) to which exemplary spectrum 130 corresponds is an edge crystal (i.e., a crystal residing along an edge of a corresponding block 20) which means much of the light generated thereby is not detected by PMTs. A spectrum corresponding to a more centrally located crystal (i.e., a crystal near the center of array 21 in Fig. 2) would have a peak energy level almost exactly at 180.

[0052] Referring again to Figs. 1, 3 and 4, during the commissioning procedure switch 76 is closed to identifier 78. At block 98, for each unit crystal, identifier 78 determines the peak energy level E_p of the corresponding commissioning spectrum 130. Once again, for the spectrum 130 in Fig. 6, the peak level is approximately 115. The peak levels are provided to determiner 80.

[0053] Determiner 80 also receives an energy target input E_t which indicates a target energy level for each crystal that corresponds to 511 keV. The target level in the present case is 180 which is provided by a system 8 user.

[0054] At step 100, determiner 80 combines the target energy level E_t with each of the separate peak energy levels E_p for each crystal thereby generating a separate gain factor G_f for each unit crystal. This combining step includes dividing the target level E_t by each of the peak levels E_p. For instance, in the case of the crystal corresponding to spectrum 130 in Fig. 6 with a peak level E_p of 115, the gain factor G_f would be 1.57 (i.e., 180/115 = 1.57). The gain factor G_f is a factor by which the energies in the commissioning spectrum for the corresponding crystal have to be shifted in order for a shifted peak energy level E_p to be equal to the target energy level E_t. Thus, in the present example, by multiplying each energy level in spectrum 130 by factor G_f = 1.57, a compensated spectrum having a desired peak at 180 results.

[0055] The gain factors G_f are stored for each separate crystal at step 102. During calibration, factors G_f are provided to shifter 82.

[0056] Referring now to Figs. 1, 3, 5, and 6, after the commissioning procedure has been completed and gain factors G_f for each crystal stored, calibration process 114 is performed to adjust compensators 70A through 70D prior to each data acquisition procedure or on a potentially repeated basis as shown by calibration process 140 within FIG. 6.

[0057]

At block 104 a radionuclide is provided within a patient for imaging purposes and the patient is positioned within area 12 adjacent detector block 20 and, specifically, adjacent the 25 block unit in the present example so that photons are directed at unit crystals during the calibration process. It has been found that when a patient who has been injected with a radionuclide is near an imaging bore, even 0.1 millicurie of

activity results in a block count rate of approximately 200 counts per second. It has also been found that approximately 5000 counts are needed in a spectrum to provide desired precision. In the present case, where the unit includes 25 blocks, for the unit, 5000 counts can be obtained in approximately one second. Thus, in the present case, the data acquisition portion of the calibration process would only require approximately one second.

[0058] Referring still to Figs. 1, 3, 5, and 6 at block 106, during a calibration period, generator 74 receives the energy E_i and crystal identifier C_i signals for every crystal within the unit and generates a separate calibration energy spectrum for each separate crystal. The calibration spectrums are similar to commissioning spectrum 130 in Fig. 7 plotting counts against energy values E_i to form the spectrum. The primary difference between the calibration and commissioning spectrums generally is that the count values will be much greater for the commissioning spectrum than for the calibration spectrum because the commissioning period (e.g., several minutes) is much longer than the calibration period (e.g., 1 or 2 seconds). During calibration switch 76 is open to identifier 78 and closed to shifter 82.

[0059] At block 108, after the calibration spectrums have been generated and stored for each unit crystal, spectrum shifter 82 receives the spectrums and the crystal specific gain factors G_f and multiplies the energy levels in the calibration spectrum by the gain factors G_f. For instance, assume that it has been some time since the commissioning procedure was performed to generate the gain factors G_f and that crystal performance has degraded somewhat. In this case, a calibration spectrum for crystal 182 may have slid such that a peak energy level for the calibration spectrum is approximately 110 (i.e., the peak has slid 5 from the peak level of the commissioning spectrum). Here, the multiplying step 108 would multiply the calibration spectrum for crystal 182 by crystal specific gain factor 1.57 thereby shifting the entire calibration spectrum and generating a shifted spectrum for the crystal.

[0060] Thus, upon shifting, the peak energy level for the shifted spectrum will be similar to value 180. In the present case, the peak level for the shifted spectrum would have a value 172.7. Other spectrum energy levels are similarly shifted by factor 1.57.

[0061] It should be appreciated that, while crystal specific data collected during a short

calibration acquisition is used to generate a crystal specific spectrum, because only a small amount of data (e.g., 2-3 hundred counts) for the crystal can be collected during the short calibration period, the crystal specific spectrum alone is not statistically very inaccurate.

[0062] At block 110 normalizer 84 receives the shifted spectrums from each unit crystal and combines all of the data corresponding to the shifted spectrums into a single normalized unit spectrum. An exemplary unit spectrum 132 is illustrated in Fig. 8 where all of the data points from the separate crystal spectrum are overlaid onto a single graph and a curve 134 is formed therefrom. In Fig. 8 it can be seen that the peak energy level of the unit spectrum 132 is approximately 175.

[0063] Continuing, at block 112, identifier 86 determines the peak unit energy level E_{up} for the unit spectrum.

[0064] At block 116, comparator 88 compares the peak unit energy value E_{pu} (e.g., 175 in the present example) with the target energy value E_t (e.g., 180) and determines the percent difference between the two values. For instance, in the present case, where the peak unit value E_{pu} is 175 and the target value is 180, the difference is approximately 2.78%.

[0065] The difference value is provided to adjuster 89 which, at block 118, adjusts the PMT gains for all of the unit PMTs via compensators 70A-70D to increase the gains as a function of the difference value. In order to avoid oscillations, adjuster will typically be set to modify the PMT gains by a percentage of the difference value. For example, an exemplary percentage may be 75% so that, where the difference value is 2.78%, the adjustments would increase gains by 2.09%.

[0066] It should be appreciated that, while insufficient counts are collected on a crystal by crystal basis to provide statistical certainty required for spectral analysis, where counts from many crystals are combined, the number of counts is sufficient to facilitate acceptable accuracy despite a short (e.g., one second) calibration acquisition period. Thus, calibration can be performed quickly, relatively accurately and without operator interaction.

[0067] Turning now to Fig. 6, a calibration procedure 140 similar to the calibration

procedure 114 is shown. In Fig. 6, however, the calibration procedure is preferably initially performed prior to image acquisition and then may be repeated such as during the image acquisition procedure. Such a continual calibration procedure provides continuous update of gain to minimize impact on patient imaging efficiency and maximize capability to maintain stable adjustment over long periods of image acquisition, which may even last an hour or more. If the sufficient activity is detected, as determined by step 120, then the calibration procedure 140 will be repeated starting at step 106. By "sufficient activity", it should be understood that the gain adjustment does require the presence of some positron activity, and therefore there should be sufficient counts to warrant a recalculation. This makes it impractical to perform the calibration procedure all day long, but instead is repeated as long as there is sufficient activity to make the update worthwhile. A specific count threshold may be set by an operator or may be preprogrammed within the processor 40. When there are sufficient counts to perform an update, when the number of counts exceeds the count threshold, the sampling would occur every few minutes and updating if there are sufficient counts observed to calculate a change and sufficient change observed to justify an adjustment. To that end, the algorithm 140 preferably detects low levels of activity and suspends updates at that time, that is, when the counts are less than the count threshold. It is possible that, even during patient acquisition, if insufficient activity is detected, as determined by step 120, then the calibration procedure 140 will end. By "end", it should be understood that the calibration procedure 140 will again be implemented prior to another image acquisition procedure.

[0068] An assembly incorporating the calibration procedure 140 will include the hardware capability to acquire energy spectra concurrently with patient coincidence data. Yet another alternative implementation may include the same hardware as shown while incorporating the ability to switch periodically and rapidly from a mode of patient acquisition to one of spectra collection and then back, although repeated and continual adjustment, as deemed necessary by a count threshold, as described within Fig. 6 is preferred.

[0069] It should be understood that the methods and apparatuses described above are only exemplary and do not limit the scope of the invention, and that various

modifications could be made by those skilled in the art that would fall under the of the invention.

[0070] To apprise the public of the scope of this invention, the following claims are made: